

TITLE OF THE INVENTION

ULTRASOUND DIAGNOSTIC APPARATUS

CROSS-REFERENCE TO RELATED APPLICATIONS

5 This application is based upon and claims the  
benefit of priority from the prior Japanese Patent  
Application No. 11-309381, filed October 29, 1999,  
the entire contents of which are incorporated herein by  
reference.

BACKGROUND OF THE INVENTION

10 The present invention relates to an ultrasound  
diagnostic apparatus suited for a contrast echo method  
using an ultrasound contrast agent mainly composed of  
microbubbles.

15 An ultrasound image diagnostic apparatus has many  
advantages which other modalities, e.g., an X-ray  
diagnostic apparatus, X-ray computer tomography  
apparatus (CT scanner), magnetic resonance imaging  
apparatus (MRI), and nuclear medicine diagnostic  
apparatus (gamma camera, SPECT, and PET), do not have.  
20 For example, the ultrasound image diagnostic apparatus  
can generate and display images almost in real time  
concurrently with scanning. This apparatus is compact  
and inexpensive. In using the apparatus, there is no  
chance of exposure to radiation. The apparatus can  
25 easily visualize blood flow.

(Contrast agent)

Ultrasound contrast media are now being put into

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practice, and an improvement in the precision of  
vascularity examination using such a contrast agent is  
expected. A contrast agent mainly consists of  
microbubbles of several microns. Microbubbles rapidly  
5 shrink and collapse upon reception of strong ultrasound  
waves. The contrast enhance effect therefore lasts  
only for a relatively short period of time.

In actual scan operation, since fresh contrast  
agent is continuously supplied to a scan slice over a  
10 blood flow, a certain degree of contrast enhance effect  
may be maintained. In general, however, ultrasound  
waves are applied several thousand times per second.  
In organs in which the blood flow rate is low or a  
relatively thin blood vessel, a contrast enhance effect  
15 can be maintained only for a moment.

(Intermittent Transmission Method)

That a contrast agent instantaneously collapses  
upon application of ultrasound waves is the most  
serious problem in the contrast echo method. As a  
20 technique of solving this problem, an intermittent  
transmission method is available. In this technique,  
scanning is intermittently repeated in synchronism with  
the R waves in an electrocardiogram. Fresh contrast  
agent continuously flows into a slice in the intervals  
25 between scans. This ensures a contrast enhance effect  
for every scan.

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(Harmonic Imaging)

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The utility of the above contrast echo method is augmented if it is used in combination with a harmonic imaging method. Microbubbles vibrate nonlinearly when ultrasound waves collide with them. This nonlinear vibrations produce harmonic components having frequencies of integer multiples of a fundamental frequency. Harmonic imaging is an imaging technique of extracting harmonic components from a fundament frequency component and visualizing them. Organs hardly vibrate nonlinearly, and hence harmonic components from the organs is pelatively small. Consequently, a region where a contrast agent exists is relatively emphasized.

As described above, if the intermittent transmission method is used, since fresh microbubbles flow into a slice during scan intervals, a contrast enhance effect can be maintained. On the other hand, in order to ensure a certain spatial resolution, about 100 to 200 ultrasound scanning lines are required per frame. That is, ultrasound pulses are applied at least a number of times equal to the number of ultrasound scanning lines in one scan.

In scanning, ultrasound pulses are transmitted while their directions are gradually changed. Therefore, microbubbles on a scanning line adjacent to a scanning line having undergone transmission/reception

may not collapse. In practice, however, since an ultrasound beam has a certain width, most microbubbles on the adjacent scanning line collapse in many cases.

In addition, the width of an ultrasound beam  
5 changes in accordance with depth. The contrast enhance effect therefore may change depending on depth. Specific problems will be described below with reference to FIGS. 1A and 1B.

When ultrasound pulses are generated from a  
10 plurality of arrayed transducers 51 while the generation timing is gradually shifted, a convergent sound field is formed. The depth of ultrasound focus point can be arbitrarily changed by changing the timing. Referring to FIG. 1A, when a focus point  
15 is formed on a scanning line with an arrow 52 at a relatively short distance, a region exhibiting a relatively high pressure sound, i.e., a region 53 exhibiting a relatively high degree of microbubble collapse (microbubble collapse region), is indicated by  
20 the hatching. The dotted lines represent adjacent ultrasound scanning lines 54.

As shown in FIG. 1A, a short-distance focus point is often formed by driving only transducers near the center instead of driving all the transducers 51.  
25 In this case, in the short-distance region, the ultrasound beam is narrow, and hence the microbubble collapse range 53 is also narrow. In the long-distance

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region, the ultrasound beam is relatively wide,  
but the ultrasound scanning line pitch is also large.  
In addition, ultrasound waves greatly attenuate because  
of the long propagation distance. As a result, the  
microbubble collapse range 53 becomes relatively  
narrow.

FIG. 1B shows a microbubble collapse range 55  
when a focus point is formed at a relatively long  
distance. As shown in FIG. 1B, to ensure a certain  
degree of sound pressure at the focus point even at  
a long distance with a long propagation distance,  
many transducers must be driven at a high voltage.  
In this case, in the short-distance region, the  
ultrasound beam becomes wide and the microbubble  
collapse range 55 also becomes relatively wide.  
As a consequence, microbubbles on adjacent ultrasound  
scanning lines collapse. When, therefore, ultrasound  
transmission/reception is performed afterward in this  
scanning line direction, microbubbles will have  
collapsed and been lost.

Another problem will be described next. Consider  
a case wherein a cardiac minor axis image is obtained  
by the contrast echo method. As shown in FIG. 2, if  
a focus point is set at a short distance in accordance  
with a depth 61 of a cardiac muscle front wall portion  
61, a rear wall portion 62 is hardly visualized due to  
biological damping and a curtain effect due to

microbubbles filling a cardiac cavity.

If a focus point is set at a long distance in accordance with the rear wall portion 62, applied ultrasound pulses collapse microbubbles on adjacent  
5 ultrasound scanning lines in the short-distance region. Therefore, no contrast enhance effect can be expected at the front wall portion 61.

It is known that in the sector scan method, when ultrasound waves are transmitted in a direction greatly  
10 shifted from a direction perpendicular to the probe vibration surface, a decrease in sound pressure and an increase in side lobe level occur as compared with application of ultrasound wave in the direction perpendicular to the problem vibration surface.  
15 That is, when ultrasound waves are applied to a cardiac muscle side wall portion 63, side lobes collapse microbubbles near the front wall portion 61 and rear wall portion 62.

It follows from the above description that it is  
20 practically impossible to transmit ultrasound waves so as to form a uniform sound field distribution on an entire scan slice. The luminance reference level is therefore not uniform within the scan slice and varies. Since scattering by tissue exhibits a relatively linear  
25 response, luminance can theoretically be made uniform by level correction for reception signals. However, a response of microbubbles exhibits strong nonlinearity,

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and phenomena such as nonlinear vibrations, expansion, and collapse of microbubbles, vary in a complicated manner depending on the absolute level of applied sound pressure. Therefore, it is practically impossible to correct reception signals.

#### BRIEF SUMMARY OF THE INVENTION

It is an object of the present invention to allow an ultrasound diagnostic apparatus to exhibit a contrast enhance effect based on an ultrasound contrast agent within a slice as uniformly as possible.

An ultrasound diagnostic apparatus according to the present invention comprises an ultrasound diagnostic apparatus for obtaining an ultrasound image of a subject into which an ultrasound contrast agent mainly composed of microbubbles is injected, comprising: a probe configured to transmit/receive an ultrasound wave to/from the subject; a transmission circuit configured to drive the probe to transmit an ultrasound wave while sequentially changing a direction of an ultrasound transmission line; a reception circuit configured to generate reception line data of the number of parallel reception from ultrasound echo signals obtained by one ultrasound wave transmission; a transmission/reception control circuit configured to control the transmission and reception circuits to change the number of parallel reception during a scan sequence for generating a 1-frame ultrasound image; and

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an image processing unit configured to generate an ultrasound image on the basis of the reception line data.

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5 Additional objects and advantages of the invention will be set forth in the description which follows, and in part will be obvious from the description, or may be learned by practice of the invention. The objects and advantages of the invention may be realized and obtained by means of the instrumentalities and combinations particularly pointed out hereinafter.

BRIEF DESCRIPTION OF THE SEVERAL VIEWS OF THE DRAWING

15 The accompanying drawings, which are incorporated in and constitute a part of the specification, illustrate presently preferred embodiments of the invention, and together with the general description given above and the detailed description of the preferred embodiments given below, serve to explain the principles of the invention.

20 FIG. 1A is a view showing a microbubble collapse range when a focus point is formed at a relatively short distance in the prior art;

FIG. 1B is a view showing a microbubble collapse range when a focus point is formed at a relatively long distance;

25 FIG. 2 is a view for additionally explaining problems in the prior art;

FIG. 3 is a block diagram showing the arrangement



of an ultrasound diagnostic apparatus according to the first embodiment of the present invention;

FIG. 4 is a view for explaining parallel signal processing used in scan operation in the first embodiment;

FIG. 5A is a view for explaining the first partial scan operation in the first embodiment;

FIG. 5B is a view for explaining the second partial scan operation in the second embodiment;

FIG. 6 is a view showing the details of FIG. 5A;

FIG. 7A is a view for explaining a multifocus method used in another partial scan operation in the first embodiment;

FIG. 7B is a view for explaining the partial scan operation in the first embodiment which uses the multifocus method in FIG. 7B;

FIG. 8 is a view showing an example in which the partial scan operation in the first embodiment is applied to a linear scan;

FIG. 9 is a block diagram showing the arrangement of an ultrasound diagnostic apparatus according to the second embodiment of the present invention;

FIGS. 10A, 10B, 10C, and 10D are views showing an example of local region segmentation in the second embodiment;

FIG. 11 is a view showing another example of local region segmentation;

FIG. 12 is a view showing another example of local region segmentation;

FIG. 13 is a view showing a scan operation sequence for local regions in the second embodiment;

5        FIGS. 14A and 14B are views showing scan operation for local regions using a parallel signal processing method in the second embodiment;

FIG. 15 is a view showing changes in focal length with movement of a transmission beam;

10        FIG. 16 is a view showing an ultrasound wave application procedure using a multishot method in the second embodiment;

FIGS. 17A, 17B, and 17C are views showing a method of displaying local regions in the second embodiment;

15        FIG. 18 is a block diagram showing an ultrasound diagnostic apparatus according to the third embodiment of the present invention;

FIG. 19 is a view showing a scanning procedure in the third embodiment;

20        FIGS. 20A, 20B, 20C, 20D, and 20E are views showing an example of a combined image of luminance portions according to the third embodiment; and

FIGS. 21A and 21B are views showing display examples in the third embodiment.

25        DETAILED DESCRIPTION OF THE INVENTION

The preferred embodiments of the present invention will be described in detail below with reference to the

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views of the accompanying drawing. Consider a clinical case wherein an anomalous region is identified by diagnosing the flow of blood into the liver parenchyma or cardiac muscle by the contrast echo method.

5 (First Embodiment)

(Arrangement and Flow of Signals)

FIG. 3 shows the arrangement of an ultrasound diagnostic apparatus according to the first embodiment. An ultrasound probe 1 is connected to an apparatus body 20. The apparatus body 20 scans the inside of an object to be examined by using an ultrasound beam through the ultrasound probe 1, creates tomographic image data by processing the obtained reception signal, and displays the image. An operating panel 9 having a trackball 10A, keyboard 10B, and the like is connected to the apparatus body 20. Various operator instructions such as an instruction to set a region of interest (ROI) are input to the apparatus body 20 through the operating panel 9.

20 A plurality of electroacoustic conversion elements (transducers) are arrayed on the distal end portion of the ultrasound probe 1. One or a few adjacent transducers constitute one channel. RF voltage pulses are applied from a transmitting unit 2 of the apparatus body 20 to the transducers. The transducers convert the electrical vibrations of the RF voltage pulse into mechanical vibrations. With this operation,

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the transducers generate ultrasound waves. Time differences in the application timing of high-frequency voltage pulses are provided between channels. These time differences, delay times in general, are set such that ultrasound waves generated by the transducers are combined into one narrow beam, and the beam is deflected as needed. By changing the delay times, the focal length and deflection angle can be arbitrarily changed. A transmission/reception control circuit (T/R) 13 controls these delay times.

Ultrasound waves propagate through the object and are reflected by an acoustic-impedance discontinuous surface located at some point in the object. The reflected waves return as echoes to the probe 1.

The echoes mechanically vibrate the transducers of the probe 1. As a consequence, weak current signals are generated. A receiving unit 3 amplifies the current signals in units of channels, converts them into voltage signals, and converts them into digital signals. In addition, the receiving unit 3 adds the signals while giving them delay times that differ between the channels. This addition is processing called digital beam forming, by which a reception signal is given a directivity. The receiving unit 3 has a plurality of digital beam forming systems. The transmission/reception control circuit (T/R) 13 simultaneously generates a plurality of reception

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signals having different directivities by parallel processing and differently controlling delay times between the digital beam forming systems.

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5 The transmission/reception control circuit 13 implements scan operation (to be described later) according to the present invention by controlling delay times in transmission and reception. Note that scan operation is defined as operation to acquire a plurality of reception signals required for the

10 creation of a 1-frame image. More specifically, scan operation is operation of repeating a series of ultrasound beam transmitting/receiving operations with respect to a plurality of ultrasound scanning lines constituting a scan plane. This new scan sequence

15 makes it possible to obtain both the effect of improving the time resolution and the effect of making the contrast enhance effect of microbubbles relatively uniform within a scan plane.

20 In addition to the transmission/reception delay control function, the transmission/reception control circuit 13 has basic control functions such as the function of setting a transmission frequency and the function of shaping the waveform of a transmission pulse. As described above, by changing the delay times

25 in transmission/reception, the transmission direction, reception direction, focal length, and ultrasound scanning line density of ultrasound beams can be

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a diagnosis or the like. In this case, the information is output to the display 7 through the B-mode DSC 5 and combination circuit 6.

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5 An ECG analyzer 12 analyzes the electrocardiograph (ECG) data of the object measured by an ECG 11, extracts, for example, R waves, and generates a trigger signal to the transmission/reception control circuit 13. The ECG analyzer 12 converts the electrocardiograph data into display data and sends it  
10 to the combination circuit 6. This electrocardiograph data and tomographic image data are combined into a single frame to be displayed together on the display 7. A clock 15 is used to control intermittent transmission intervals in a diagnosis using on ECG  
15 signal, e.g., a diagnosis of an abdominal organ. Note that the operator can control the intermittent transmission intervals and timing on the operating panel, and the control operation is reflected in the transmission/reception control circuit 13.

20 (Scan Operation)

The scan operation in this embodiment will be briefly described by exemplifying the parallel signal processing of simultaneously generating a plurality of reception signals having different directivities for  
25 one transmitting operation. Consider a case wherein two reception signals having different directivities are obtained for one scanning operation. FIG. 4 is

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a view for explaining the principle of parallel signal processing. Referring to FIG. 4, "r" represents an ultrasound scanning line; and "t", the transmission direction. Ultrasound pulses are transmitted under  
5 delay control corresponding to the direction of an ultrasound scanning line t1. From the resultant echo signals, digital beam formers of two systems generate two types of reception signals that are given directivities in the directions of ultrasound  
10 scanning lines r1 and r2 under two types of delay control. A reception scanning line density twice a transmission scanning line density is realized by this parallel signal processing. Obviously, four or more reception signals with different  
15 directivities can be theoretically generated for one transmitting operation.

The scan operation in this embodiment will be described in detail next with reference to FIGS. 5A and 5B. Assume that one scan plane consists of 160  
20 ultrasound scanning lines. The angle difference between adjacent ultrasound scanning lines is represented by  $\theta$ .

Scan operation for acquiring reception signals required to generate a 1-frame image is constituted by  
25 a plurality of, two in this case, partial scan operations. More specifically, in the first partial scan operation, the first portion in the scan region is



scanned, and in the second partial scan operation,  
the remaining, second portion in the scan region is  
scanned. A partial image of the first portion obtained  
by the first partial scan operation and a partial image  
5 of the second portion obtained by the second partial  
scan operation are combined into one frame, thereby  
completing a 1-frame image of the entire scan plane.

In the first and second partial scan operations,  
the focus point is fixed to a long distance. FIG. 5A  
10 shows the first partial scan operation. FIG. 5B shows  
the second partial operation. As shown in FIG. 5A,  
ultrasound transmission/reception is repeated in  
predetermined cycles. A transmission beam is  
sequentially moved from the right end to the left  
15 end of a scan plane in each transmission/reception.  
The intervals at which a transmission beam is moved are  
set to an integer multiple equal to or more than two of  
the angle difference  $\theta$  between ultrasound scanning  
lines,  $4 \cdot \theta$  in this case.

20 In digital beam forming, a plurality of, four  
in this case, reception signals having different  
directivities are generated in every transmission by  
parallel signal processing. More specifically, four  
reception signals having directivities corresponding to  
25 fourth ultrasound scanning lines which are symmetrical  
about a transmission beam are generated.

These four reception signals are subjected

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to detection and luminance conversion in the receiver 4  
and written in the B-mode DSC 5. Different  
write sequences for these signals are used in  
a short-distance region A and long-distance region B.  
5 In the short-distance region A, all the four reception  
signals are written. In the long-distance region B,  
only the two reception signal corresponding to the two  
ultrasound scanning line located in the center are  
written.

10 This first partial scan operation will be  
described in detail below with reference to FIG. 6.  
FIG. 6 shows a microbubble collapse range 100 in which  
a high contrast enhance effect is obtained when the  
focus point is formed at a relatively long distance.  
15 The long-distance focal point is formed under a typical  
transmission condition, i.e., a wide aperture and high  
driving voltage. In this condition, the width of the  
microbubble collapse range 100 is wide at the  
short-distance region A and narrow at the long-distance  
20 region B, as described with reference to FIG. 1B. The  
contrast enhance effect is high within the microbubble  
collapse range 100.

In accordance with this region exhibiting a high  
contrast enhance effect, in the short-distance region  
25 A, all four reception signals (luminance signals)  
corresponding to four ultrasound scanning lines r1, r2,  
r3, and r4 which are symmetrical about a transmission

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beam t11 are written in the B-mode DSC 5. In the long-distance region B, only the two reception signal corresponding to the two ultrasound scanning lines located in the center are written in the B-mode DSC 5.

5           The transmission beam is then moved to t12, and all four reception signals (luminance signals) corresponding to four ultrasound scanning lines r5, r6, r7, and r8 which are symmetrical about the transmission beam t12 are written in the B-mode DSC 5. In the  
10           long-distance region B, only the two reception signals corresponding to the two ultrasound scanning lines located in the center are written in the B-mode DSC 5.

          When the first partial scan operation is completed  
15           upon repeating this sequence, on the memory of the B-mode DSC 5, the entire short-distance region A is filled with the luminance data, and the long-distance region B partially has luminance data blank portions. These blank portions are filled with the luminance data  
20           obtained by the second partial scan operation.

          Disregarding averaging, the number of times of transmission/reception required for the first partial scan operation is  $160/4 = 40$ .

          As is obvious, since only the reception signals  
25           within the microbubble collapse range 100 need be finally used for an image, unnecessary reception signals need not be generated in the first partial scan

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5           The second partial scan operation will be  
described next. In the first partial scan operation,  
data blank portions are present at two adjacent  
ultrasound scanning lines in the long-distance  
region B. Referring to FIG. 5A, for example,  
0           ultrasound scanning lines r4 and r5 correspond to  
blank portions. In the second partial scan operation,  
transmission, digital beam forming, and DSC write  
operation are performed to fill these blanks with data,  
as shown in FIG. 5B.

15           Ultrasound transmission/reception is repeated in  
predetermined cycles like the first partial scan  
operation. A transmission beam is sequentially moved  
from the right end to the left end of a scan plane in  
every transmitting/receiving operation. The intervals  
20   at which the transmission beam is moved are set to  $4 \cdot \theta$   
as in the first partial scan operation. However, the  
transmission beam in the second partial scan operation  
is shifted from the transmission beam in the first  
partial scan operation by half the moving intervals of  
25   the transmission beam, i.e.,  $2 \cdot \theta$ . With this  
operation, an ultrasound beam is transmitted in the  
direction of the center axis of each ultrasound

scanning line corresponding to a blank portion formed  
in the first partial scan operation. The first  
transmission beam in the second partial scan operation  
is transmitted to a direction  $t_{21}$  between blank  
5 scanning lines  $r_4$  and  $r_5$ . The next transmission beam  
is transmitted in a direction  $t_{22}$  between blank  
scanning lines  $r_8$  and  $r_9$ . In this manner, transmission  
is repeated at intervals of  $4 \cdot \theta$ .

In digital beam forming, a plurality of, two in  
10 this case, reception signals having different  
directivities are generated by parallel signal  
processing for each transmitting operation.  
More specifically, two reception signals having  
directivities corresponding to left and right  
15 ultrasound scanning lines on the two sides of  
a transmission beam are generated.

These two reception signals are subjected to  
detection and luminance conversion in the receiver 4,  
and only the data corresponding to the data blank  
20 portions in the long-distance region B formed in the  
first partial scan operation are written in the B-mode  
DSC 5.

As described above, in the second partial scan  
operation, image portions are assembled such that they  
25 do not overlap the image portions generated in the  
first partial scan operation in FIG. 5A, and no blank  
portions are formed. The number of times of

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transmission/reception in FIG. 5B is set to 40 (39, to be exact) as in the case shown in FIG. 5A. That is, transmission/reception is performed a total of 80 times, which is equal to the number of times of transmission/reception in 2-direction parallel signal processing. That is, the frame rate does not decrease.

The merits of this scan operation will be described below. At a long-distance focus point, the beam width increases in the short-distance region A. In this short-distance region A, microbubbles collapse in a wide range. In other words, in the short-distance region A, a contrast enhance effect owing to microbubbles can be obtained in a wide range. This makes it possible to effectively generate reception signals at once from a wide range corresponding to as many as four scanning lines for one transmitting operation by 4-direction parallel signal processing. On the other hand, in the long-distance region B, the intervals between ultrasound scanning lines become larger than those between ultrasound scanning lines in the short-distance region A, and the distance from the center of a transmission beam increases, resulting in a decrease in the intensity of ultrasound waves. If 4-direction parallel signal processing is performed in this state, sensitivity deteriorates. For this reason, in the first partial scan operation, 2-direction parallel signal processing is performed.

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In the second partial scan operation, then, only the blank regions in the first partial scan operation are filled with data. That is, the blank regions in the first partial scan operation correspond to data on scanning line pairs in the long-distance region. At a long-distance focus point, the scanning lines in these blank regions are included in the microbubble collapse range exhibiting a high contrast enhance effect. In this manner, high contrast enhance effects can be ensured in both the long- and short-distance regions.

In this case, for the sake of simplicity, the short-distance region A and long-distance region B are separated from each other by a clear boundary. To obtain a smoother image, these regions may overlap. In this case, for example, the luminance image on the overlapping portion is averaged to make the boundary less noticeable.

This scan operation is especially effective for sector scan operation in which ultrasound scanning line intervals differ in a short-distance region and long-distance region. However, effects similar to those described above can also be obtained when this operation is applied to linear scan operation in which ultrasound scanning line intervals remain unchanged in a short-distance region and long-distance region.

In addition, this technique may be used in

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combination with the intermittent transmission method of performing transmission in synchronism with an ECG signal. In this case, since the application of ultrasound waves is stopped during periods other than synchronous periods, more microbubbles flow into a slice of a region of interest and are stored without collapse. Obviously, if ultrasound waves are applied in this state, more microbubbles can be detected. In addition, if the scanning method of this embodiment is used, the contrast enhance effect can be relatively improved.

(Other Examples of Scanning Method)

A multifocus method is available as a conventional method similar to the scanning method to be described below. The multifocus method is currently implemented in many apparatuses. This method will be described first with reference to FIGS. 7A and 7B. According to the multifocus method, as shown in FIG. 7A, (a) the respective ultrasound scanning lines are scanned to perform transmission/reception at a short-distance focus point so as to generate a signal component corresponding to a short-distance region A, and (b) a signal component corresponding to a long-distance region B is generated at a long-distance focus point. These two images are then combined into a 1-frame tomographic image. Each ultrasound scanning line has two focus points (represented by the heads of



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moves backward in the transmission direction at each set. In addition, focus points are alternately switched in the short-distance region and long-distance region in each transmission, thereby obtaining the following effects.

Assume that the ultrasound scanning lines are sequentially scanned from the ultrasound scanning line r1 as in a conventional method. In this case, when an ultrasound pulse is transmitted to the ultrasound scanning line r1, microbubbles on the ultrasound scanning line r2 are affected, e.g., collapse. However, since the ultrasound scanning line r2 is scanned first, this adverse effect can be avoided. In addition, since the ultrasound scanning line r2 is scanned at a short-distance focus point, the influence of this scanning on the next ultrasound scanning line r1 is small. That is, microbubbles on the ultrasound scanning line r1 do not collapse much. As described above, this method can minimize the collapse of microbubbles on adjacent ultrasound scanning lines due to the application of ultrasound waves.

Note that if different focus points are set on the respective ultrasound scanning lines as in this case, since different sound fields are formed, echo signals on adjacent ultrasound scanning lines may be made uneven. Smoothing by, for example, averaging on adjacent ultrasound scanning line to reduce such

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unevenness is effective for an improvement in image quality.

Although the scanning method of the above embodiment is a scanning method represented by a sector probe, the present invention can also be applied to a linear type scanning method. As shown in FIG. 8, in the linear type scanning method, the intervals between ultrasound scanning lines do not depend on depth.

However, the profile of the sound field formed by one beam is the same as that in the above method, and the influences on microbubbles on adjacent ultrasound scanning lines in a short-distance region still remain.

When the present invention is to be applied to the linear type scanning method, ultrasound scanning lines are formed to be spaced from each other by a distance  $4d$ , and the ultrasound scanning lines are shifted by a distance  $2d$  to form ultrasound scanning lines spaced part from each other by the distance  $4d$ . In this case, the time required to generate one frame becomes

equal to that in the scanning method in which the transmission ultrasound scanning line density is  $2d$ .

This embodiment presents a scanning procedure by which signals derived from microbubbles can be efficiently received, and the unevenness of contrast enhance effect within a slice can be corrected when one tomographic image is to be generated by the contrast echo method performed by administering a contrast

agent. With this procedure, even if an ultrasound contrast agent exhibiting the same performance as that of a conventional contrast agent is administered, the contrast enhance effect can be relatively improved.

5 Therefore, an improvement in blood flow diagnosis ability, especially an improvement in fine blood flow diagnosis ability, is expected.

(Second Embodiment)

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10 The second embodiment provides a partial imaging method. The partial imaging method is a method of segmenting a scan plane into a plurality of local portions and scanning each local portion in an optimal scan operation sequence, instead of sequentially moving over scanning lines within the scan plane, thereby  
15 obtaining optimal (maximum) contrast on the entire scan plane and combining the resultant data into a one frame.

(Arrangement and Flow of Signals)

20 FIG. 9 is a block diagram showing an ultrasound diagnosis apparatus of this embodiment. The same reference numerals as in the first embodiment denote the same parts in the first embodiment, and a detailed description thereof will be omitted. A template memory  
21 stores pieces of information about a plurality of  
25 models for segmenting a scan slice into a plurality of local regions (partial regions). One optimal pattern for a sliceal shape of a portion to be diagnosed is

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First of all, it is important to set a specific local portion toward which the diagnostic apparatus is

to perform optimal scanning. The settings vary depending on an organ to be diagnosed and its slice. For example, the cardiac muscle in a cardiac minor axis image has an almost circular shape. A circular  
5 template 50 representing a local region like the one shown in FIG. 10A is selected in advance. The size of the template 50 is then adjusted by using the zoom function of the operating panel 9 or the like to almost overlap the outer ring of the template 50 on the minor  
10 axis image, as shown in FIG. 10B. If the size of a template is fixed in accordance with a routine, the template need not be displayed in some case.

As shown in FIG. 10C, the operator then designates desired local regions with points or regions.

15 Obviously, a plurality of regions can be designated. In this case, regions can be set at uneven intervals along the cardiac muscle. As shown in FIG. 10D, the operator designates representative points of those set above. With this operation, the apparatus  
20 performs automatic segmentation. In this example, diametrically located four points are designated to obtain a segmented region like the one shown in FIG. 10B. Obviously, the size of a template, the number of local regions segmented, and the like  
25 can be changed.

In the above case, the cardiac minor axis image is exemplified. However, a template suited for the shape

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or the like of another slice, such as a 2-cavity cross section or major axis image, may be selectively used. In addition, the setting method shown in FIGS. 10C and 10D can also be used. In diagnosing the liver, since the liver is included in an overall slice, relatively simple local region segmentation like the one shown in FIG. 11 can be performed.

(b) Scan Operation

The set local regions are sequentially scanned.

In general, this scanning is started when the operator presses a start button on the operating panel 9.

In the case shown in FIG. 12, for example, when a local region A is to be irradiated, a focus point is set at the central portion of the region A.

A transmission/reception control circuit 13 changes the ultrasound transmission conditions in accordance with the position of each local position (transmission focus point) so as to almost equalize the degrees of dynamic influences on the respective local regions, i.e., sound pressures on the respective local regions and the degrees of collapse of microbubbles in the respective local regions. Typical transmission conditions that can be adjusted include the driving voltage for each transducer, the aperture width (the number of transducers to be simultaneously driven), the driving frequency, and the scanning line density.

In this case, the respective parameters are

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changed to almost equalize the degrees of collapse of  
microbubbles in consideration of biological damping  
(mainly determined by the transmission distance of  
ultrasound waves) and the irradiation angle with  
5 respect to each transducer. When the parameters  
associated with the driving voltage for each transducer  
and the number of transducers to be simultaneously  
driven are changed, the degrees of collapse of  
microbubbles can be made almost uniform by almost  
10 equalizing the sound pressures at the positions of the  
respective focus points. More specifically, when the  
driving voltage for each transducer is to be changed,  
the driving voltage is lowered if the focus point is  
located near, and vice versa. When the number of  
15 transducers to be driven is to be changed, the number  
of transducers to be driven is decreased if the focus  
point is located near, and vice versa. When the  
frequency is to be changed, the frequency is increased  
if the focus point is located near, and vice versa.  
20 Note that only one of the above parameters may be  
changed or a plurality of parameters may be  
simultaneously changed.

Strictly speaking, the value of damping varies  
among objects to be examined. However, a rough value  
25 can be presented to the operator on the basis of data  
obtained in advance by measurement. With reference to  
a region E exhibiting the maximum damping, relative

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transmission sound pressures for the respective regions are set such that -1.5 dB is set for regions D and F; -2 dB, for regions C and G, -4.5 dB, for shallow regions B and H, and -6 dB, for a shallowest region A.

5           If an effective echo signal is to be obtained by irradiating the local region E with ultrasound waves, the region A on the same ultrasound scanning lines is affected by the collapse of microbubbles. It is therefore useless to scan the region A immediately  
10           after the region E. This problem can be effectively solved by using an intermittent transmission method.

          As shown in FIG. 13, for a circulatory organ region, transmission is performed once for every heartbeat or for a few heartbeats by using triggers  
15           synchronous with an electrocardiogram. In the interval between triggers, no transmission is performed, and hence new microbubbles flow into a region of interest upon collapse of microbubbles. This makes it possible to acquire a sufficient contrast enhance effect again.

20           In the case shown in FIG. 13, at the same trigger timing, echo signals are generated from the pairs of local regions B and H, C and G, and D and F at the same heartbeat timing by using the parallel signal processing method. This is because each pair of  
25           regions are located at the same depth and same focal length, and do not affect each other, i.e., do not collapse microbubbles each other, since they are spaced

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apart from each other. In addition, in the case shown in FIG. 13, scanning from the region A to the region H is repeated a plurality of number of times.

FIGS. 14A and 14B show another operation procedure. Assume that scanning is started from the right end of the screen. In this case, at the first trigger, as shown in FIG. 14A, the local regions A, B and C are simultaneously scanned by using 3-direction parallel signal processing. In each region, however, the focal length and output sound pressure can be changed, and the changed values are set as optimal values in each region. At the second trigger, as shown in FIG. 14B, similar scanning is performed for regions on which the ultrasound scanning lines at the first trigger overlap. In this method, the segmentation forms of local regions are not uniform unlike those in FIG. 11. A merit of the method is that microbubble echoes can be obtained relatively effectively with a smaller number of times of transmission/reception.

FIG. 15 shows an advanced method. At the first trigger, the focal length changes for each ultrasound scanning line (an illustration of ultrasound scanning lines is omitted), as indicated by "A" in FIG. 15. The output sound pressure is also controlled in accordance with a change in focal length such that output pressures on other ultrasound scanning line in this focus point portion are made uniform. At the

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present invention described above, since data (signal intensity) presenting a contrast enhance effect on the entire region of interest can be obtained by executing at least two scan procedures, the technique of combining the data and displaying the resultant image is used. In the case shown in FIG. 12 or 14, since the boundaries between the respective local regions are clear, a simple image (simple signal intensity distribution) is reconstructed by the combination circuit 6 using only data from a corresponding region, and the image is displayed on the display 7.

This image combining and displaying method can be applied as follows. In diagnosing a myocardial blood flow, attention is often paid to find an ischemic region of the cardiac muscle. In this case, the operator need only know fine blood flow perfusion due to a contrast agent without paying any attention to the fine form of speckle pattern of the cardiac muscle (as in the case of a scintigram in nuclear medicine). To meet such needs in diagnosis, for example, a display method of calculating the average contrast degree of each local region and displaying the calculated value as the representative value of each local region is used. Painting each region by using color information based on a color bar or the like makes the displayed information easier to discern (FIG. 17A). Alternatively, simplified display may be performed as

shown in FIG. 17B. Furthermore, as shown in FIG. 17C, blood flow rates may be numerically expressed and displayed. Although the luminances and numerical values in display represent relative information, such

5 a display method allows the operator to quickly detect an ischemic region of the cardiac muscle if it exists. These images may be displayed side by side as well as being overlaid on an original diagnostic image.

More simplified display images can be added to

10 a patient's chart or electronic patient's chart through a means such as a network as well as being output from a printer.

(Third Embodiment)

FIG. 18 shows the arrangement of the third

15 embodiment. A transmission/reception control circuit 13 controls the timing of pulses from a transmitting unit. This embodiment performs intermittent transmission using a timing signal from an ECG analyzer 12 or clock 15 in accordance with a mode

20 switching instruction sent from an operating panel 9. Intermittent transmission is transmission in which frame generation intervals are sufficiently larger than those in normal continuous transmission (20 to 100 frames/sec). For example, time intervals

25 corresponding to four or five heartbeats are input.

The transmission/reception control circuit 13 instructs a transmitting unit 2 to perform transmission

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of a plurality of frames per trigger (in other words, continuous scanning in a short period of time).

FIG. 19 shows a conceptual rendering of this transmission. Referring to FIG. 19, five frames are continuously transmitted/received at predetermined time intervals (sufficiently longer than the frame rate). An image processing unit 31 combines a plurality of frames obtained per trigger by the above method and sends the resultant images to the combination circuit, and the images are displayed on the display. Prior to a description of an image processing method, diagnostic images expected from the transmission method in FIG. 19 will be described.

#### <Problems to Be Solved>

If the contrast agent concentration is relatively low or a contrast agent is made of microbubbles that easily collapse, most microbubbles on a slice collapse by a transmission pulse in the first frame. In the second and subsequent frames, therefore, diagnostic images are formed without any microbubbles, i.e., made of only tissue echoes.

It is, however, empirically known that there are cases other than the above case. If, for example, the contrast agent concentration is high or a recently developed contrast agent containing microbubbles that are relatively resistant to ultrasounds is used, the following phenomenon occurs.

As in FIG. 20A, in transmission of the first frame (A), a contrast enhance effect is seen at a relatively short-distant portion. However, the collapsing effect of high-concentration microbubbles at the short-distance portion increases, and hence ultrasound pulses can hardly propagate to a deeper portion. As a consequence, no image is seen at a portion deeper than the short-distance portion, and the portion becomes dark. In some case, such a portion becomes darker after administration of the contrast agent than before (called shadowing). As shown in FIG. 20B, in the second frame (B), since the microbubbles at the short-distance portion have collapsed, the contrast enhance effect decreases. However, since damping of sound waves due to microbubbles is reduced, a relatively deep portion is irradiated with a relatively high sound pressure. The contrast enhance effect at this portion increases. Subsequently, a similar phenomenon is transferred to deeper portions. When microbubbles collapse in all regions as shown in FIG. 20C, the displayed image is formed by tissue echoes (FIGS. 20D and 20E).

If this phenomenon is seen with a moving picture, the movement of the luminance based on contrast from a shallow portion to a deep portion like a curtain that drops is recognized. This phenomenon will be referred to as a "curtain phenomenon" hereinafter.

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Obviously, when the above phenomenon occurs, the operator cannot examine a contrast enhance effect on an entire slice by only seeing one of a plurality of images. All the images must be joined to each other.

5 <Arithmetic Image Processing>

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The image processing unit 31 described above performs optimal image combining processing when the above curtain phenomenon occurs. More specifically, images obtained at one trigger timing are stored in  
10 the image memory 8. The luminance signals of these images at the same coordinates in the respective frames are compared with each other to detect maximum values. Arithmetic processing for determining a luminance  $I(x, y)$  at coordinates  $(x, y)$  is given by

15 
$$I(x, y) = \text{MAX}(I_i(x, y)), \quad I = 1 \cdots N$$

where  $I_i(x, y)$  is the luminance at the coordinates  $(x, y)$  in the  $i$ th frame, and  $N$  is the number of images to be compared with each other.

As a result of this processing, images like those  
20 shown in FIGS. 20A, 20B, 20C, 20D, and 20E having high-luminance contrast portions joined together are combined, and the combined image is displayed on the display. As is obvious from the result, since this image displays the luminance corresponding to the  
25 highest contrast among all the regions, the operator can examine the overall contrast degree with this one diagnostic image.



Note that this technique is similar to an MIP (Maximum Intensity Projection) method used to project three-dimensional space information on a two-dimensional plane. However, the general MIP method is used for spatial points, whereas the technique of the present invention is used for temporal points.

Note that the above arithmetic processing is relatively simple, and a combined image is preferably displayed almost in real time immediately after transmission at the trigger timing.

As described above, this embodiment cannot exhibit a sufficient effect when no curtain phenomenon occurs, but has no adverse effect. Therefore, this technique is not used in any specified condition.

#### <Examples of Display>

FIGS. 21A and 21B show examples of a display form. FIG. 21A shows a method based on two-window display. While intermittent transmission is observed in real time on one window, the above combined image is sequentially displayed on the other window. Referring to FIG. 21B, images obtained at one trigger timing are displayed side by side, and a combined image is simultaneously displayed. Note that all image need not have the same size. In general, since a combined image is most important for diagnosis, the image is preferably displayed in a relatively large size, as shown in FIG. 20B.

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Additional advantages and modifications will readily occur to those skilled in the art. Therefore, the invention in its broader aspects is not limited to the specific details and representative embodiments shown and described herein. Accordingly, various modifications may be made without departing from the spirit or scope of the general inventive concept as defined by the appended claims and their equivalents.

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